Design of novel ventricular chambers: comparison of three different models

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***Abstract* — Mechanical circulatory supports have to be carefully designed with the aim of increasing hemocompatibility and avoiding blood stagnation to prevent thromboembolic events. Explicit FEM analysis of three different designs was preliminary performed by means of LS-DYNA to assess pros and cons of each model. A hemocompatible polycarbonate urethane was chosen to simulate material's mechanical characteristics.**

***Keywords*— ventricular chamber, FEM analysis, design, LS-DYNA.**

I. Introduction

TOTAL ARTIFICIAL HEARTS, also known as TAHs, are mechanical circulatory supports that replace biological hearts in case of refractory end-stage heart failure [1]. TAHs are mostly applied as bridge-to-transplantation devices, but they are currently receiving a growing interest as destination-therapy solutions; indeed, many drawback are still limiting their exploitation (i.e. dimensions, power supply, cardiac output, …).

Nowadays, CardioWest TAH represents the only device that already received the FDA approval and CE mark: up to date, it has been implanted in more than 1300 patients [1]. Another promising device is the Carmat TAH, which is equipped with highly sophisticated electronics [2]. These two TAHs share the same working principle: their ventricular chambers are made of a rigid dome and a flexible membrane; the back and forth motion of the membrane produces systolic and diastolic effects, making the blood flow possible. This design assures the pulsatile flow, but many researchers are currently evaluating the possibility to use non-pulsatile pumps, i.e. rotary ones. These latter come from the success of rotary Left Ventricular Assist Devices (LVADs). The possibility to sustain life without a physiologic blood flow was confirmed by Golding et al. in 1980 [3], but the long-term usage of a continuous flow pump can lead to many collateral risks, such as microvascular dysfunction and haemorrhages [4].

Through a preliminary FEM analysis, the present work investigates three different ventricular chamber's geometries as possible innovative solutions for a novel pulsatile device. The simulated material is a polycarbonate urethane (PCU), well known for its bio- and hemocompatibility.

The computational analysis aimed at examining the tensile state of stresses and deformations generated in the material during one systolic and diastolic cycle: too high stresses or strains can result in the early failure of the device.

The simulations showed how to improve the design of the pulsatile chamber.

II. Materials and Methods

A. 3D models

Three different linear geometries were developed starting from an initial concept of a bellow-like pump [5]. The original sketch shows an ellipsoidal flexible chamber provided with two one-way heart valves. Compression and expansion of the chamber (thanks to an electromagnetic actuator) operates the motion of the chamber; valves assure the correct blood flow. 3D real-scale models were created using AutoDesk Fusion 360.

The first model has a three-sphere-shaped geometry (Figure 1): two one-way heart valves (21 mm) are located inside the necks between the central chamber and the lateral ones. The second model (Figure 2) was designed to avoid excessive radial expansion when the central chamber reaches maximum axial compression. In this case, valves (31 mm) are placed at the level of the two rigid plates. The third model (Figure 3) is aimed at eliminating the rigid plates: this is possible thanks to a pair of specifically–designed valves (50 mm). In this configuration, valves are able to compress the rib in the middle, working as the plates work in the previous models. Thus, this geometry does not produce excessive radial expansion.

Fig.1: First model: two rigid plates compress the central compartment

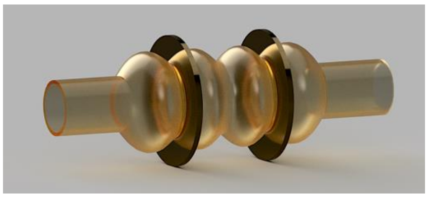

Fig. 2: Second model: it avoids excessive radial expansion

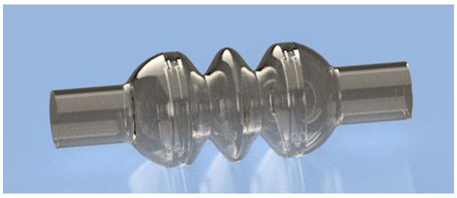

Fig. 3: Third model: the use of rigid plates is not necessary

B. Element formulation

The geometry of each chamber was imported in LS-DYNA Pre-Post. Each chamber was approximated as a thin-wall object and shell elements (2 mm thick) were used. The plates were idealized as rigid bodies: a solid elements mesh was used for these components.

The prototype's material was described trough *MAT_MOONEY-RIVLIN_RUBBER model (Material Type 27) to simulate the rubber-like behaviour of the PCU. The rigid plates were described with *MAT_RIGID (Material Type 20), inputting the properties of Aluminium (i.e. Young's Modulus, Poisson's ratio and density).

C. PCU mechanical properties

The mechanical characterization of PCU (ChronoFlex AR-LT, AdvanSource Biomaterials, Wilmington, MA, US) was fundamental to get the data that were used in the material's cards (data not shown). Uniaxial tensile test was performed (with an elongation rate of 1 mm/s) using dog bone shaped sample. The acquired stress-strain engineering curve was uploaded in LS-DYNA. The Poisson's ratio was also measured: during the tensile test thickness, width and gage length were taken at different elongations using a digital caliber. The average value of the Poisson's ratio was used in the material's card in LS-DYNA. A density of 1200 Kg/m^3 was considered.

D. Contacts and constrains

Contacts are necessary to avoid elements penetration: to this purpose AUTOMATIC_SINGLE_SURFACE and AUTOMATIC_SURFACE_TO_SURFACE were defined for preventing penetration between shell elements, and shell and solid elements, respectively.

In the models provided with the rigid plates, another contact was defined: TIED_NODES_TO_ SURFACE. It allowed to stick the prototype's nodes set (corresponding to the valve's region, this set is the slave part) to one of the two plates (master part); this contact was useful for avoiding any kind of sliding between the ventricular chamber and the plates.

Translation and rotation of the nodes/elements of the cylindrical regions were constrained in all directions: in fact, these parts will be fixed to the chassis of the device. Moreover, the rigid plates/valve locations are free to move back and forth only following the pump axial direction. Constrains are represented in Figure 4.

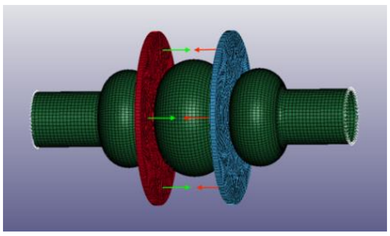

Fig. 4: Constrains are applied to the first model: to the cylindrical extremities (white rings) and to the moving plates (green and red arrows). Plates can get back to the original position)

E. Actuation

In order to mimic the actuation of each prototype, the symmetric motion of the rigid plates (or of the valves) was defined: a displacement vs. time curve was associated to the moving components using BOUNDARY_PRESCRIBED_ MOTION. A preliminary assessment was aimed at estimating the capacity of the prototypes to withstand the imposed displacement of the valves/rigid plates after one cycle. The critical state of stress and highest strains were identified to check if they are low enough for a possible fatigue analysis. The study was carried out without considering fluid dynamics.

The movement of the plates/valves is very fast: they need to go forth and back at least 2 times per second to achieve a minimum flowrate of 5 L/min. The prototypes have an internal capacity from 60 mL to 70 mL and the internal fluid is supposed to be not completely ejected.

III. RESULTS

A. Stresses and strains

The three-sphere-shaped prototype (Figure 5) reached less than 1 MPa stress when the central chamber is completely flattened, with a maximum engineering strain of 18.3%. These values are low, but the diameter of the central compartment largely increased during compression: from 66 to nearly 80 mm. For this reason, this model was discarded.

The second prototype (Figure 6) developed 0.9 MPa and a strain of 19.5% in the necking region between the two rigid plates. This model negligibly increased its maximum diameter (from 55 to 57 mm) and it will be taken into account for further investigations.

The last model (Figure 7) did not appreciably change its diameter (from 60 to 65 mm), but it showed much higher stresses and strains with respect to the first two prototypes: 3.5 MPa with an engineering strain of 60%. These values were reached because of the design of the lateral compartments: the motion of the two valves caused an intensive stretch in these regions during the chamber's compression. However, stresses and strain can be reduced by modifying the geometry of these regions.

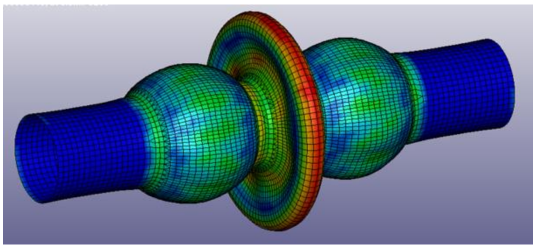

Fig. 5: First model: effective stresses when the ventricular chamber is flattened (the rigid plates are not shown): maximum values are in red

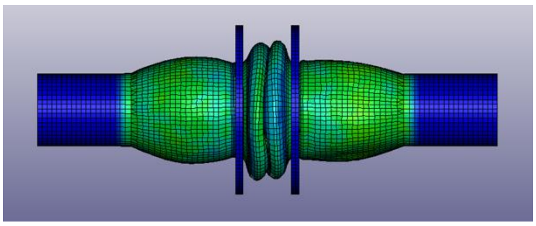

Fig. 6: Second model at the maximum compression: stresses are shown. The highest values are in the middle of the two ribs

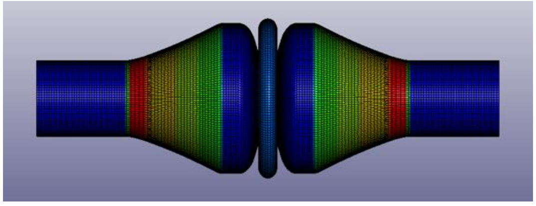

Fig. 7: Third model at the maximal compression: stresses of 3.5 MPa are in red

IV. CONCLUSION

This preliminary computational analysis of three different ventricular chambers allowed identifying the most suitable geometries. In particular, the second model assures a negligible increase of diameter and low stresses; the third model has to be re-designed to reduce stresses and deformation.

Further analyses are needed, in particular fatigue analyses and fluid dynamics simulations, before moving to the production of a real prototype for the functional tests.

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